# Speech-perception aids for hearing-impaired people: Current status and needed research

Working Group on Communication Aids for the Hearing-Impaired[a)]
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Both the overall aging of the population and its exposure to higher noise levels have increased the tendency to hearing loss and the importance of improved hearing aids for speech perception. This article reviews improvements in conventional electroacoustic hearing aids, as well as recently developed alternative classes of speech-perception aids, including surgically implanted cochlear stimulators, and vibrotactile, electrocutaneous and optical stimulating devices. It is concluded that the most effective aid for the vast majority of hearing-impaired persons is, and will remain for the immediate future, the electroacoustic hearing aid. In those cases for which no benefit is demonstrated for the electroacoustic aid, generally meaning persons with profound hearing loss, either the cochlear implant or a tactile aid may provide some improvement in the understanding of speech. In rare cases, some speech understanding in the absence of lip reading is achieved by patients with cochlear implants, for unexplained reasons. This and other pressing questions about speech processing need to be addressed by the research community if more effective aids are to be developed for the use of the 36.5 million hearing-impaired persons expected in the U.S. by the year 2050.

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[a)] Please see the end of this report for an important notice and the members of the Working Group on Communication Aids for the Hearing-Impaired and of the Committee on Hearing, Bioacoustics, and Biomechanics (CHABA).



  

TABLE I. Estimated proportion of the U.S. population with hearing impairment (in millions), 1983.

| Year | U.S. population | Number with hearing impairments | Percentage hearing-impaired |
|---|---|---|---|
| 1960 | 180.6 | 13.0 | 7.2% |
| 1970 | 205.0 | 15.2 | 7.4% |
| 1980 | 227.7 | 18.1 | 7.9% |
| 1990 | 249.7 | 21.1 | 8.4% |
| 2000 | 268.0 | 24.0 | 9.0% |
| 2025 | 301.0 | 32.7 | 10.9% |
| 2050 | 308.9 | 36.5 | 11.8% |

1989). Over half of the hearing-impaired people in the United States were working-age adults, and roughly one-third were over 65. These data, based on survey interviews, do not indicate the levels of severity of the hearing impairments. In 1985 about 8% of the U.S. population (about 19 million people) had a hearing impairment; roughly 3% (about 7 213 000 people) indicated that they could understand normal conversation only "with difficulty;" and 0.3% (about 481 000 people) said they could not hear normal conversation at all.

The primary treatment for most hearing-impaired people, namely those who have sensorineural hearing loss ("nerve deafness") and cannot or do not choose to learn to live with it, has been some type of aid for speech perception. Most commonly these have been conventional electroacoustic hearing aids. However, modern technology has not neglected this problem, and a variety of new types of aids has been developed. Some of these have been commercially available for 10 to 15 years, while others have only recently been shown to be effective in laboratory studies and have just begun to be used in significant numbers. Some of these new devices have been the subject of a good deal of discussion in the general media, including, for example, reports of President Reagan's in-the-canal acoustic hearing aids.

For the severely impaired and deaf population, there is no device that has caught the imagination of the public more than the cochlear implant or aural prosthesis. This device has been developed on the basis of animal work by Galambos and Rupert (1959), Simmons (1964), and others and applied to human patients by Simmons (1966), House (1974), Michelson (1971), and by many others in recent years. Newspaper headlines such as "New Ear Implant Liberates the Deaf from World of Silence" (*Indianapolis Star*, 30 January 1985) have announced this new device to the public. When computerlike memory banks are available in $30 wristwatches, the idea that a replacement for the organ of hearing might be devised and surgically implanted seems not at all farfetched, at least to some people. However, as is often the case, enthusiastic reports in the general media have failed to reflect a wider range of opinions in the clinical and scientific literature. For example:

> Results have been very encouraging, with all but two of these patients [out of 15 implanted and tested with a UCSF/Storz multichannel cochlear implant] obtaining some degree of open-set auditory-only speech recognition. Most patients have demonstrated improvement over time without extensive rehabilitative intervention and all patients have attained an enhancement in lip-reading ability, as measured with a tracking procedure. According to a self-rated performance scale, all patients have experienced improvement in general communicative function since receiving the implant (Schindler and Kessler, 1987).
>
> Although it is true that several implant recipients have reported achieving open-set speech discrimination, it is equally true that the auditory skills of some implant recipients do not exceed the skill level they achieved with a hearing aid. In fact, most persons who are involved in the clinical investigation of cochlear implants likely would be quick to point out that superior performance is probably the exception rather than the rule (Windmill *et al.*, 1987).

While the attention of the medical community, as well as the general public, was understandably captured by the concept of a "bionic" cochlear implant, research has also continued on other approaches to the treatment of hearing impairment. In addition to a wide range of technological improvements in traditional acoustic hearing aids, significant advances have also been made in sensory-substitution aids. Rather than eliciting responses in the cochlea and VIIIth cranial nerves with intense sound (as in the hearing aid) or by direct electrical stimulation (as in the cochlear implant), this approach simply bypasses the auditory sections of the nervous system and presents speech information via an unimpaired sensory channel (vision or touch). That speech can be received through the tactile channel is demonstrated by the abilities of some deaf-blind individuals who communicate by tactile speechreading (the TADOMA method). These people are able to understand speech at slow conversational rates by placing a hand on the face of the talker and monitoring the mechanical actions of the face associated with speech production. An equivalent demon-

TABLE II. Age distribution and prevalence rates of hearing-impaired people in the United States, 1988. (Note: Total hearing-impaired population figures in Table I (1983) disagree with those shown in Table II for 1988, because of discrepancies used in projections for the former. The resulting errors do not exceed 2%–3% overall.)

| Age | Hearing-impaired people (millions) | Prevalence rate (%)[a] |
|---|---|---|
| Under 18 | 1.1 | 1.7 |
| 18–44 | 5.0 | 4.9 |
| 45–64 | 6.7 | 14.8 |
| 65–74 | 4.8 | 27.4 |
| 75 and over | 4.2 | 38.1 |
| All ages | 21.9 | 9.1 |

[a] Ratio of number of cases divided by number of people in the total population in that age group. Source: National Center for Health Statistics data cited in Adams and Hardy (1989).

are, in some programs, first fitted with a vibrotactile device. Unusually good performance with the sensory-substitution aid is seen by at least one implant team as an indication that a patient will also do well with the implant and therefore that the implant procedure should be undertaken. Others find such decisions difficult to understand, in the absence of a larger volume of evidence that the implant, on the average, can be predicted to provide better speech perception than a vibrotactile device.

This report is not intended as a substitute for the thorough reading of the pertinent literature, which is the responsibility of any clinician or scientist who wishes to achieve expertise on speech-perception aids. It does attempt, however, to articulate both the views of a representative sample of those who have studied these subjects actively for the past 20 to 30 years, as well as something of the variance in opinions about optimal treatment held by contemporary clinicians and scientists (despite their general agreement about the results of relevant research).

It is hoped that the report will provide some useful assistance to several groups who may lack either the time or the appropriate background to study carefully the literature on speech-perception aids for hearing-impaired and deaf people. One important group is deaf and hearing-impaired people with the technical background necessary to read it. Although we wish that this report could promise more help for those people than it does, a careful reading should assist them in making difficult choices. No one is likely to be a more dedicated advocate for a patient than the patient. A second target group is physicians in general practice and other clinicians who may not be actively involved in selecting or providing hearing-impaired people with speech-perception aids but who nevertheless have to advise hearing-impaired patients. Third, it is hoped that this report may be useful to the staff of, and advisers to, the federal agencies responsible for the support of research on the next generation of communication aids. Finally, hearing-aid manufacturers may find this report useful in their consideration of future product lines.

## I. CONVENTIONAL ELECTROACOUSTIC HEARING AIDS

In this section, we first describe the characteristics of electroacoustic hearing aids that are currently available.[1] We then discuss some of the major results of research efforts to improve electroacoustic aids, many of which have not yet been incorporated into commercially available aids. Finally, we discuss the characteristics of those hearing-impaired individuals for whom traditional electroacoustic aids are the most appropriate aid to speech perception.

Acoustic amplification is the method most commonly used to enhance the recognizability of speech and other information-bearing signals so as to improve communication for hearing-impaired people. The instrument most commonly used for this purpose is the conventional hearing aid. A conventional hearing aid has three key characteristics: (1) it is a personally fitted, wearable device; (2) it can be freely put on and taken off by the user, although intended for regular full-time use; and (3) the amplified signal is delivered acoustically to the external ear canal. Other acoustic amplification devices that do not meet these constraints are known more generally as "assistive listening devices;" this class of devices is not discussed here.

Conventional hearing aids typically consist of a microphone, electronic filter, controls for adjusting the amplification (or gain) and overall shape of the frequency response (e.g., a bass or treble boost), circuits for limiting the amplified signals to a comfortable or safe level, an earphone (commonly referred to as a "receiver" in the United States), a battery that serves as the power source, and various acoustic components, such as flexible tubing and an earmold, for coupling the output of the receiver to the external ear canal.

### A. Introduction to conventional electroacoustic hearing aids

#### *1. Major types currently available*

In this section, we describe the major types of conventional electroacoustic hearing aids currently available. Discussions of the frequency of use of each of the types and their relative merits appear in later sections. Because of the wide variety of types of conventional aid and because merits or drawbacks often characterize more than one type, this organization seemed most efficient.

A common method of classifying conventional hearing aids is according to their relative size and how they are worn. The largest wearable hearing aid is the body aid, in which the electronic components are housed in a body-worn case, the amplified signals being delivered by wire to a receiver mounted in the ear. Most body aids are high-powered instruments.

A smaller and less conspicuous instrument is the eyeglass hearing aid. The electronic components of this device fit into one of the bows of a pair of eyeglasses. The microphone is mounted at the front of the bow next to the lens, the receiver is at the rear of the bow and its output is via a nozzle and attached tubing to an earmold.

A relatively small and very popular instrument is the behind-the-ear (BTE) hearing aid. The electronic components of a BTE aid are housed in a small elliptical case that, as the name implies, fits behind the ear. The acoustic signals generated by the receiver are delivered to the ear canal by means of a flexible acoustic tube terminating in an earmold.

An even smaller instrument is the in-the-ear (ITE) hearing aid. All the components of this aid are contained in a small plastic case that is molded to fit into the user's ear. It typically occupies the outer portion of the external ear canal and the concha (the innermost part of the external ear).

The smallest hearing aid of all is the in-the-canal (ITC) aid, which fits entirely in the ear canal. It is the least conspicuous of conventional hearing aids, the outer face of the unit being just visible at the opening to the ear canal. Unfortunately, the available power decreases with size, due primarily to the very small batteries required for these instruments and the limited electrical power they supply. As a result, ITC instruments are generally restricted in application to those patients with only mild amounts of hearing loss.

Hearing aids are also classified according to the number

cuous units that fit either behind or in the ear. Some body-worn aids are still used for special applications. Although outside the scope of this report, there are other commercially produced acoustic amplification systems for hearing-impaired people. These include special-purpose amplifiers for use on the telephone, radio, and television as well as personal FM or infrared transmission systems for use in classrooms, auditoriums, theaters, and other settings in which background noise or reverberation is a problem.

### *3. A recent development: Implantable electroacoustic aids*

A recent development, used to date in relatively few patients, is implantable hearing devices (not to be confused with cochlear implants). These devices typically do not involve an air-conduction path but rather substitute direct mechanical stimulation for the amplified acoustical signal. There is at least one major advantage of direct stimulation of the middle ear—reduction in acoustic feedback—and various techniques for direct mechanical stimulation have been developed over the years. One approach is to attach a small magnet to the eardrum or at some other point in the ossicular chain and to drive the magnet electromagnetically by means of an induction coil (Watanabe, 1965; Glorig *et al.*, 1972).

Another approach is to insert a metal pin into the temporal bone and to drive the pin electromagnetically (or by an external vibrator), the vibrations being transmitted to the cochlea by bone conduction (Hough *et al.*, 1986). An implantable hearing device using this general approach is available in which a small magnet is implanted in the mastoid. The magnet is sealed in silicone and housed in a titanium disk attached to an orthopedic screw. The screw-magnet assembly, referred to as the internal unit, can be implanted under local anesthetic in an outpatient setting.

The internal unit is driven by an induction coil mounted externally. The magnetic core of the coil is used to hold the external unit in place directly over the implanted magnet. Electrical signals applied to the coil cause the magnet to vibrate. These vibrations are transmitted to the cochlea by bone conduction. The implantable bone-conduction hearing device has been designed primarily for persons with conductive hearing impairment and for whom surgical correction is inappropriate. The device has recently received marketing approval from the Food and Drug Administration and as of 1 May 1987, more than 120 persons had been implanted with this device.

## B. Dealing with reduced dynamic range and related audiological considerations

As is evident from the preceding section, cosmetic and marketing factors have had a major influence on hearing-aid development. As a consequence, engineering efforts have focused primarily on issues of microminiaturization and power conservation (for reducing battery size). Audiological considerations, despite their importance, have remained of secondary concern to the industry. This section provides a review of the basic audiological constraints affecting hearing-aid performance and the various attempts to deal with these constraints.

The reduced dynamic range of the impaired auditory system is the most obvious audiological problem and has received the greatest attention. The nature of the problem is illustrated in Fig. 1. The solid curves show typical speech spectra as measured at a distance of 1 m from the speaker's lips. The dashed curves show hearing thresholds for three typical sensorineural impairments. (Sensorineural impairment has been chosen for this illustrative example since it is by far the most common impairment for which hearing aids are prescribed; in this form of hearing impairment, greater losses are typically found in the high frequencies.) Curve A is for a person with a mild-to-moderate, high-frequency hearing loss, curve B is for a severe loss, and curve C is for a profound hearing loss. Curve D shows the normal threshold of hearing. All curves show the detectability of a 1/3 octave band of noise as a function of band center frequency. The dashed line at the top of the diagram is the loudness discomfort level curve. Loudness discomfort levels are fairly similar for both normal hearing and sensorineurally hearing-impaired persons and, for purposes of simplicity, a single curve is shown.

The effective dynamic range of the auditory system is defined by the distance between the threshold curve and the loudness discomfort level (LDL) curve. The area encompassed by these two curves, i.e., between the threshold of audible sound and the ceiling of too-loud sound, is known as the residual hearing area. Note that the residual hearing area becomes progressively smaller with increasing hearing loss. Under extreme conditions, particularly in the high frequencies, the dynamic range from threshold to discomfort level may be only a few decibels. Some hearing-impaired persons with a drastically reduced dynamic range will typically expe-

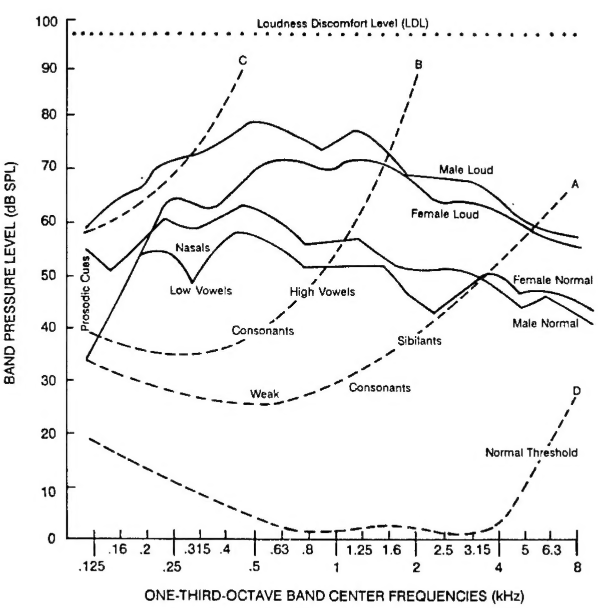


FIG. 1. Illustrations of dynamic range of the impaired auditory system.

abilities of *average* hearing-impaired listeners (Pavlovic, 1984; Kamm *et al.*, 1985; Humes *et al.*, 1986). To illustrate expected speech-recognition performance, let us consider the three hearing-impaired listeners shown previously in Fig. 1. Articulation Index calculations were made for quiet listening conditions for each of these listeners. For these calculations, we have assumed that a representative prescriptive procedure is used (gain equal to one-half the hearing loss) and that the maximum real-ear gain realized is 50 dB at any frequency. We have further assumed a normal conversational speech level (70 dB SPL) and speech materials consisting of nonsense syllables (Resnick *et al.*, 1975). The unaided and aided speech-recognition performance estimated for patients A, B, and C under these assumptions are as follows:

| Patient | Unaided | Aided |
|---|---|---|
| A (mild-to-moderate) | 79% | 93% |
| B (severe) | 41% | 66% |
| C (profound) | 0% | 12% |

Clearly, as the severity of loss progresses, the aided speech-recognition performance decreases. This is true even though the amplification of the hypothetical instrument is considerably greater for the profoundly impaired listeners. Estimates of aided performance for meaningful sentences rather than nonsense syllables are 100%, 91%, and 15% for patients A, B, and C, respectively. All estimates of speech-recognition performance for these patients, moreover, are for auditory input only; visual cues are not considered. It is important to note again that, although these articulation index predictions have been shown to be accurate for average listeners with a given hearing loss, some differences remain for individuals with identical loss.

### 3. Protecting against excessive amplification

In addition to frequency shaping, it is also necessary to provide some form of protection against excessive amplification. Amplification is excessive when it either results in sound levels that are uncomfortably loud for the patient or causes further loss of hearing due to the high sound levels. It is inevitable, however, that hearing aids for severely impaired listeners will generate high sound levels in the listener's ear and could jeopardize their remaining hearing sensitivity. General clinical practice is to assume that, although the high gain required by severe-to-profound losses for optimal speech recognition may continue to do some damage to the remaining sensory and neural structures, this risk of treatment-induced loss is preferable to the social isolation resulting from the nonuse of amplification. It is also assumed, from the fairly modest changes in threshold that typically occur in patients with severe or profound hearing loss following use of high-gain aids, that the remaining healthy sensory and neural elements in such an ear are fairly resistant to intense stimulation (Humes and Bess, 1981). More research, however, is needed on this issue.

Two of the most common forms of protection against excessive amplification are peak clipping (eliminating all portions of the output of the aid that exceed some specified level) and compression limiting (reducing amplification for higher-level sounds). The latter form of protection produces less distortion. Compression amplification, in principle, can also be used to maximize the proportion of the time-varying speech spectrum that can be placed within the residual hearing area. Clinical protocols for the specification of hearing-aid output levels not exceeding loudness discomfort have been developed in recent years (Hawkins, 1980; Cox, 1981), as have guidelines for limiting the output to safe levels that minimize the risk of further loss of hearing (Humes and Bess, 1981).

### 4. Compression amplification

Methods of compression amplification can be divided into three broad categories: (1) compression limiting, (2) long-term automatic gain control, and (3) syllabic compression. Each of the above methods of compression can be implemented either within a single frequency band (wideband compression) or in several contiguous frequency bands (multiband compression). Unless otherwise stated, wideband compression is assumed in the discussion that follows.

The most common form of compression in modern hearing aids is that of compression limiting. This type of compression is designed primarily for protection and operates only at relatively intense sound levels. The hearing aid behaves as a conventional amplifier for signals below the threshold of compression. When the compression threshold is exceeded, which occurs only for signals approaching the discomfort level, the gain of the amplifier is reduced substantially so that the output does not exceed a hazardous or uncomfortable sound level.

Experimental evaluations have shown that, as a protective device, compression limiting is superior to simple peak clipping. There is less distortion of the amplified speech signal and, correspondingly, speech intelligibility is reduced less by compression limiting than by peak clipping (Davis *et al.*, 1947). Clinical evaluations of compression limiting, however, have not been favorable (Blegvad, 1974; Edgardh, 1952). In a critical review of this topic, Braida *et al.* (1979) attribute the negative results obtained in clinical evaluations to poor choice of compression characteristics, lack of individualized fitting, and confounding with other, uncontrolled electroacoustic variables.

Long-term automatic gain control (reducing amplification in response to high output levels), also known as automatic volume control (AVC), involves the use of a relatively long time constant; i.e., a time constant much greater than the duration of individual syllables in speech. This form of compression is designed to adjust for long-term variations in speech level so that more of the speech signal lies within the available range of residual hearing. Although the potential value of long-term automatic gain control was recognized some time ago, few studies have been undertaken to evaluate this form of compression. In addition, for reasons that are unclear, only a small proportion of conventional hearing aids have long-term automatic gain control.

In syllabic compression, the parameters of the amplification system are chosen so as to alter the relative intensities of individual speech sounds. Many of these intensity changes

more scientific work has been done with them. Second, as noted in the final conclusions and elsewhere in this report, the conventional electroacoustic aid will almost certainly be the treatment of choice for most hearing-impaired people for the foreseeable future, and thus research on those devices is aimed at a far larger population of prospective users. Third—and probably most important—many of the special signal-processing strategies currently being investigated for use in electroacoustic aids will, once perfected, very likely be applicable to both of the other types of aid (cochlear implants or sensory-substitution aids).

### *1. Digital hearing aids*

A very promising new development is that of the digital hearing aid. Digital techniques are now commonly used in instruments for hearing-aid measurement and calibration. As a result, powerful new measuring tools have been developed. These include instruments for the rapid and convenient measurement of sound transmission in the ear canal, measurement of *in situ* gain of hearing aids, and more accurate measurement of acoustic impedance. The extension of digital technology to the hearing aid itself appears to be primarily a matter of time and several experimental digital hearing aids have already been developed (Levitt, 1982, 1987; Nunley *et al.*, 1983; Engebretson *et al.*, 1986a, 1987; Cummins and Hecox, 1987).

Three types of digital hearing aids have been developed: (1) a quasidigital hearing aid in which digital circuitry is used to control analog amplifiers and filters; (2) a sampled-data system in which the audio signal is sampled at discrete intervals in time, the samples remaining in analog form during processing; and (3) the all-digital hearing aid in which the audio signal is sampled, converted to binary form, and then reconverted back to a continuous analog waveform after processing.

Current research is focused on developing application-specific chips that are both small enough and have sufficiently low power consumption to provide a viable alternative to the conventional BTE hearing aid. Several leading industrial research laboratories are actively engaged in the development of digital hearing aids. A practical compromise that appears to be feasible for the immediate future is that of a quasidigital hearing aid in which the audio signals remain in analog form but are controlled by digital circuitry. A chip for adaptive noise reduction in hearing aids has been developed using this approach (Graupe *et al.*, 1987).

Digital hearing aids promise many advantages over conventional analog instruments (Levitt, 1987). In addition to providing greater accuracy and flexibility in the choice of electroacoustic parameters, they can be programmed by an external computer, thereby allowing for the introduction of new and more effective approaches to prescriptive fitting and evaluation of hearing aids (Popelka and Engebretson, 1983; Engebretson *et al.*, 1986b). Moreover, powerful new signal-processing techniques can be used for reducing acoustic feedback (Preves *et al.*, 1986) as well as enhancing speech intelligibility and reducing the effects of background noise and reverberation (Lim, 1983), as discussed shortly. It is unlikely that many of these features will be available in the first generation of commercially produced digital hearing aids, but once a practical, wearable digital amplification system has been developed, all of the above-mentioned features represent viable short-term goals.

### *2. Speech-analyzing hearing aids*

A hearing aid incorporating some degree of speech-specific signal processing is, by definition, a speech-analyzing hearing aid. Conventional hearing aids that shape the speech spectrum so as to be comfortably loud at all frequencies are thus a very simple form of speech-analyzing hearing aids. Of particular interest in this section are experimental speech-analyzing hearing aids or, more generally, methods of speech analysis that could be usefully incorporated into such aids.

*a. Frequency-lowering techniques.* A broad class of (experimental) speech-analyzing hearing aids involves a process known as frequency lowering. One of the most common characteristics of a sensorineural hearing impairment is that hearing loss is greater at higher frequencies. In particular, hearing loss due to aging (presbycusis) is characteristic of this type. Translating the acoustic spectrum of speech downward (to lower frequencies) is thus an appealing prospect, since it would transfer speech energy in the high frequencies (where it is either not audible or poorly resolved) into the low-frequency region, where it is both audible and resolvable.

A number of frequency-lowering systems have been developed over the years using various strategies to accomplish the changes, using both selective and total-waveform lowering (see Braida *et al.*, 1979, for a review of many different systems). Despite the number of frequency-lowering devices that have been invented or reinvented, a crucial question remains: How effective are these devices in improving the intelligibility of speech for hearing-impaired persons?

Experimental evaluations of frequency lowering for hearing-impaired people have yielded mixed results. Guttman and van Bergeijk (1959) reported that some improvement in speech reception could be obtained using a channel VOCODER for frequency lowering, but that learning to achieve a substantial level of improvement would be slow and time consuming. In an early evaluation of his transposer system, Johannson (1966) obtained improvements in the discrimination of fricatives and other phonemes by profoundly hearing-impaired children. In contrast, Ling (1969) reported a series of experiments in which no significant advantages over conventional amplification were obtained for the Johannson-type frequency transposer for either speech reception or speech training. Foust and Gengel (1973), however, showed significant improvements in speech discrimination ability (relative to conventional amplification) for individual subjects, but only after a fair amount of training. Reed *et al.* (1983, 1985b) have studied pitch-invariant frequency lowering, using nonuniform frequency compression of the short-term spectral envelope, but without any large-scale improvement in speech discrimination performance.

A special form of speech-feature transposition is that in which the fundamental frequency of the voice is recoded so

within the available range of residual hearing, there may be inherent cognitive limitations in learning the new code. It has been argued, for example, that the negative results obtained with most speech transposition systems are due to the transposed speech code's being too complex to be learned, even though all of the speech information may be available in the coded signal. An alternative view is that the time course of auditory perceptual learning is so long that few laboratory experiments have revealed the extent of the average human listener's ability to learn new codes (Watson, 1980).

As mentioned in the Introduction, the question whether hearing-impaired persons can learn to recognize radically distorted speech signals raises several very basic issues regarding learning, plasticity, and the information capacity of the impaired auditory system. One point of view is that, if a radically different speech code is to be learned, then this learning should occur during the early years of life, when a child is first learning the sounds of speech. The assumptions underlying this approach are very difficult to test, and there are substantial ethical implications in doing so. It may well be that a deaf child could acquire speech and language more readily if all of the important speech cues were transposed to lie within the child's region of residual hearing. If, however, the experiment is a failure, the child's acquisition of speech and language may be retarded even further than would otherwise be the case.

A counterargument to the above approach is that the auditory system may have inherent speech-feature detectors. The ability (of a child or an adult) to learn to recognize radical transformations that did not code speech with these features might therefore be severely restricted. The thrust of this argument is that acoustically normal speech is in some sense special and that the normal auditory system is uniquely structured for the processing of that form of information. Accordingly, any major deviation from the normal speech code might be extremely difficult to learn and, from this view, it would be questionable whether rates of communication comparable to that of normal speech could ever be achieved using such codes. Essentially the same argument can be made regarding the difficulties encountered in learning to communicate using visual and tactile representations of speech. While this argument may be of some theoretical interest, the success of highly skilled lipreaders and of deaf-blind users of the TADOMA method of speech perception appears to contradict a strict (acoustic) speech-is-special argument, as does the performance of some implanted children.

In commenting on an earlier draft of this report, Liberman pointed out yet another version of the speech-is-special argument. He has stressed that the perceptual unit may be the *speech gesture*, by which is meant neuromotor-control sequences normally associated with the production of phonemes or other speech units. From this view, any transformation of speech—acoustic, visual, tactile or electrical—might be successful as long as it preserved these gestures in perceptually salient forms. The abilities of excellent lipreaders, deaf-blind users of TADOMA, and some outstanding cochlear-implant patients Liberman considers consistent with this view, because each of those modes, *unlike* telegraphy or printed text, does maintain the integrity of the "code" (speech gestures). Liberman has developed this basic position, originally termed the "motor theory of speech," in numerous articles over the past three decades (e.g., Liberman *et al.*, 1967; Liberman and Mattingly, 1985). If groups of congenitally deaf children were eventually taught to communicate using a variety of different transforms of speech, some of which preserve the speech gestures and some that do not, the validity of Liberman's and other "speech-is-special" arguments might be tested in a way that was never before possible. Meanwhile, the practice of choosing transformations that are as perceptually similar to speech as possible will probably continue, and it seems likely that none of the major speech theorists will find this objectionable.

### 3. Signal processing to reduce noise and reverberation

One of the most common complaints made by hearing-aid users is that speech in noise, or speech in a reverberant room, is particularly difficult to understand. Poor speech reception in noise and/or reverberation is to be expected since a reduced set of speech cues is available to hearing-impaired persons. They are therefore less able to make use of the normal redundancy in speech to compensate for the information degraded by the noise or reverberation. This problem is not unique to hearing aids and applies equally well to cochlear implants and to tactile and visual sensory aids.

For the common situation in which speaker-to-microphone distance cannot be controlled effectively, a major reduction in the level of the background noise or of the reverberation is very difficult to achieve. There are, however, several available techniques by which small improvements can be obtained. One method is the use of a directional microphone. Many modern hearing aids use directional microphones that provide several decibels of improvement in speech-to-noise ratio over omnidirectional microphones. User reaction to directional microphones in hearing aids has been mixed. Some find directional microphones very helpful, others do not. It should be noted that directional microphones are not effective in a highly reverberant room, and also that, for those conditions under which a directional microphone works well, it is necessary to point the microphone steadily toward the speaker, which is not always a convenient maneuver.

A second improvement, which can be implemented fairly simply, is to use two microphones. The typical approach is to place one microphone on each ear. In a true binaural hearing aid, the output of each microphone is amplified and delivered to the corresponding (ipsilateral) ear. In a quasibinaural aid, the outputs of the microphones are simply added, amplified, and routed to one, or possibly both, ears in parallel (Harris, 1965).

One of the major advantages of the two-microphone systems is that, for any given frequency, at any given instant in time, the speech-to-noise ratio at one ear will be greater than that at the opposite ear. Which ear has the larger speech-to-noise ratio will depend on the relative spatial locations of the speech and noise, the acoustic shadow produced by head diffraction, and the relative spectra of the speech and noise. Simply adding the two microphone outputs will improve the

cally designed for hearing-impaired persons has been developed by Graupe and Causey (1975). Preliminary evaluations of this self-adaptive noise filter show significant improvements in speech intelligibility (Stein and Dempsey-Hart, 1984). These studies, however, did not use time-invariant filtering as a reference condition, and therefore it is not known how much of the improvement is due to the adaptive characteristics of the filter and how much to simple filtering. Tyler (1988) describes very little improvement with this type of noise filter, or with a passive reduction in low-frequency gain. Similar findings have been reported recently by Van Tasell *et al.* (1988) and Klein (1989).

The technique developed by Graupe and Causey has been implemented on a single chip (Graupe *et al.*, 1987), and several manufacturers have recently introduced noise-reducing hearing aids into their product line using this chip. Consumer reaction to this form of noise reduction has yet to be assessed.

*h. Spectrum subtraction.* Another approach to noise reduction is that of spectrum subtraction (Lim, 1978; Boll, 1979). According to this procedure, a running estimate of the noise spectrum is obtained, typically during pauses in the speech. The estimated noise spectrum is then subtracted from the short-term speech-plus-noise spectrum yielding a spectrum with a much reduced noise component. All of the above spectra are necessarily short term, i.e., time windows of finite duration are used. Since the procedure is designed to track time-varying changes in the short-term speech spectrum, these time windows should be shorter than syllabic durations in speech. Lim and Oppenheim (1979) have shown that, under certain conditions involving linear transformations of the signals, the short-term spectrum subtraction method reduces to that of adaptive Wiener filtering.

A key element in the spectrum subtraction method, as in the Wiener-filtering approach, is the accuracy with which the speech-plus-noise and noise-only spectra can be estimated. The mechanism used for deciding that noise only is present is thus of critical importance; the more accurate this decision, the more reliable is the subsequent noise reduction. Empirical evaluations of the spectrum subtraction method have generally shown improvements in speech-to-noise ratio, but no significant improvements in speech intelligibility, although improvements in overall quality have been reported (Lim and Oppenheim, 1979).

The subtraction process need not be restricted to the frequency domain. Furthermore, in addition to spectral transformations such as the Fourier transform, other nonlinear transformations can be used. In the approach used by Weiss *et al.* (1974), the square root of the amplitude spectrum is transformed to the time domain, by analogy with cepstrum analysis (Noll, 1967), and after nonlinear weighting (determined empirically) the subtraction process takes place. This method of noise reduction has yielded good results in terms of overall speech quality and in reducing fatigue when listening to speech in noise for long periods of time. Data obtained on hearing-impaired subjects show similar results (Levitt *et al.*, 1986).

*i. Comb filters: Speech-specifying filtering.* In principle, the more that is known about the signal, the more effectively it can be extracted from the noise. A number of noise reduction techniques have focused on known features of the speech signal. One class of such methods utilizes the quasi-periodic structure of the speech waveform. If the voice pitch is known, a comb filter can be used to extract all of the harmonically related components of the speech signal and to reject the noise between these harmonics. The effectiveness of this approach, however, is critically dependent on the accuracy with which the fundamental frequency of the voice has been estimated. This creates an inherent problem in that in order to reject the noise effectively a comb filter with high-frequency resolution is needed, but this requires a filtering time that is fairly long, which, in turn, may not be able to effectively track temporal variations in the voice pitch.

In one implementation of the comb-filter technique (Lim *et al.*, 1978), it was found that increasing the filter length improved the speech-to-noise ratio but reduced intelligibility. The longest filter length used was equal to 13 pitch periods of the speech. This produced an improvement of 9.8 dB in the speech-to-noise ratio, but intelligibility was reduced to less than half of that for the unprocessed condition. The shortest filter length used was equal to 3 pitch periods. This produced only a 3.4-dB reduction in speech-to-noise ratio, and no significant change in speech intelligibility. On the positive side, subjects reported that the quality of the processed signals was superior to that of the unprocessed speech in noise. [Note: Acoustical engineers typically attempt to process waveforms to maximize either speech *quality*, or *speech-to-noise ratio* (S/N), possibly because *intelligibility* is often at or near 100% in the systems under study. In the case of hearing-impaired listeners, however, the most important measure of performance is clearly the accuracy with which a listener can identify a message that was spoken.]

A case of special interest is that in which the interference is another speech signal. Parsons (1976) developed a method of harmonic selection for this particular problem. Although intelligibility data are not provided, Parsons reports that "suppression of the unwanted talker is virtually complete, except in a few cases where shared peaks produce some residual crosstalk."

*j. Speech-model-based filtering.* A second class of speech-specific procedures is concerned with modeling the speech production process and then estimating the parameters of this model. This approach can be used for both bandwidth reduction and noise reduction. Evaluations of this approach have been confined largely to the accuracy with which the model parameters can be estimated under noisy conditions. Although data show reliable estimation of certain speech parameters under noisy conditions (Lim and Oppenheim, 1979; Kobatake *et al.*, 1978; Wise *et al.*, 1976), no data have yet been reported showing an improvement in intelligibility of the reconstructed speech signal.

*k. Phoneme-specific filtering.* A third approach focuses on the acoustic structure of different classes of speech sounds. Drucker (1968) proposed a noise-reduction technique in which different filters are used for each of five classes of speech sounds (stop, fricative, glide, vowel, nasal). Each filter is designed to emphasize the salient characteris-

with average hearing loss less than 25 dB may require the use of amplification. The negative impact of mild hearing loss, in the range 15–25 dB, and of unilateral hearing loss on the educational development of children has received much attention in recent years (Bess, 1985, 1986; Oyler *et al.*, 1987).

As noted in Sec. IV, these recommendations, while typical of most practitioners, are not universally accepted, particularly by those advocates of cochlear implants who favor early intracochlear implantation of profoundly deaf children.

#### *1. The effect of age*

Age is an important variable when considering the application of prosthetic devices. In the case of young children, educators generally believe that (conventional) hearing aids should be applied as soon as possible. This includes children as young as a few months of age (Rodel, 1985; Krantz, 1985). It is not so clear whether aids should be applied in the case of a mild loss in a child under 3, although it is certainly not categorically ruled out. Few such losses are identified at this early age, in any case, and the stresses associated with keeping an aid on such a young child are considerable for both the child and the parent. Training the parent in more effective communication skills is probably a better choice in most situations.

For the somewhat older child, hearing aids are more likely to be applied successfully. This together with evidence that even mild hearing losses can significantly affect language and learning skills of school-age children (Bess, 1985; Davis *et al.*, 1986) supports the need for prosthetic devices for virtually all hearing-impaired persons in this age category.

The candidates for prosthetic devices other than conventional hearing aids are found mainly among the profoundly hearing impaired. (This point is discussed in more detail in the next two chapters.) However, in the case of very young children (birth to 2–3 years), there are two considerations that appear to preclude some options: (1) it is not possible to determine the hearing status of a child in this age range with as much certainty as in the case of older persons and (2) because these children are young, they will live a long time and during that period revolutionary new developments will probably occur. Therefore, procedures should not be undertaken that cause permanent detrimental changes in the auditory system either directly or through a lack of utilization.

In the case of many older individuals, special considerations include a general inability to cope with amplification devices. For this reason very elderly persons with mild losses may not profit enough from amplification to offset the difficulties of using it. The same may be true of the very old with very severe losses. In general, however, each person is very much an individual case for whom the best course of action depends on a variety of variables. At this time, there is no method of aid selection that is clearly more effective than a trial period with an aid that is well selected in relation to the hearing loss and to the personal needs and capabilities of the individual.

#### *2. The effect of hearing loss on speech intelligibility*

As discussed in earlier sections, conventional hearing aids make an acoustic signal more intense overall, make it relatively more intense in some frequency regions than others, limit its maximum intensity level, and reduce its intensity range. Furthermore, signals arriving from some directions may be reduced relative to those arriving from another. However, for all that, the basic nature of the signals is unchanged from the original. That is, it is a time-varying acoustic waveform that encodes, among other things, those features most listeners recognize as speech. For persons with mild to moderate hearing losses, and even for some with severe losses, this simple alteration of the stimulus processing is very much preferred to more drastic recoding because it retains the essential temporal and spectral features of the normal speech signal. Preservation of a relatively familiar signal also obviates the need for lengthy, difficult, and often only partially successful training of the user to understand a new information encoding scheme. For these reasons and because those with mild-to-moderate losses often hear quite well with conventional aids, they are the aids of choice for these groups.

With greater degrees of hearing loss, several factors combine to reduce the effectiveness of the conventional hearing aid. A greater hearing loss requires greater amplification to achieve audibility. This means that both the hearing aid and the ear itself may produce more distortion. Upward spread of masking may also be a problem at high sound levels. Furthermore, because the threshold of discomfort does not increase as much as the threshold of detection, the dynamic range of the ear can be very much reduced. For these reasons, a hearing aid must reduce the signal's dynamic range, sometimes sharply, if the hearing loss is great. As a result, the features that contribute to intelligibility are reduced further. Finally, in cases of severe-to-profound hearing loss, damaged or missing peripheral auditory-system components may be incapable of performing preliminary analyses of the signal that are necessary if the properties of the speech waveform are to be successfully communicated, via the auditory nerve, to the higher components of the auditory nervous system.

The combined effects described above may render many persons with profound hearing loss unable to derive substantial benefit from the acoustic signals provided by conventional hearing aids. Even persons with severe hearing losses may receive only very limited help from conventional hearing aids, but the amount of help provided still may be greater than that available by any other means at present. This may be true even for some persons with profound hearing losses. When a hearing loss is only moderate-to-severe in extent, it is almost certainly true that a conventional aid will provide better results than any current alternatives.

It follows from this discussion that the candidates for alternative prosthetic devices to the conventional aid are found among those with the greatest hearing losses, for the reason that those are the persons who process speech poorly, or not at all, with conventional devices. The basic question is which technology will provide the best results for each individual. As alternative methods of aiding the hearing im-

(Note: At the time this report was submitted for publication in this *Journal*, single-channel cochlear implants were no longer available in the U.S. However service remains available for previously implanted single-channel devices.)

## A. Basic psychophysics of the electrical stimulation of hearing

This review provides an overview of basic psychophysical measures collected from patients with several different single- and multichannel cochlear-implant devices. Detailed descriptions of the implants are provided in the articles cited. In general they were intracochlear implants, with the stimulating electrodes placed about 15 to 25 mm into the basal turn of the scala tympani. Single-channel stimulation is generally accomplished with the shorter electrodes, multichannel with longer. Whenever possible we have chosen results from subjects with direct percutaneous connections to their electrodes in order to avoid possible transmitter/receiver artifacts. There is, however, no reason to believe that there are any substantial differences in the responses of subjects for similar signals actually reaching the electrode, whether or not the signal is internally demodulated by a totally implanted stimulating system. Some reported results, as noted in the text, have favored the best observations rather than the average or worst. The intent of much of the published literature on cochlear implants has been to characterize the sensations that can be elicited by electrical stimulation of the peripheral auditory system, rather than to document the successes or failures with all implanted patients.

### *1. Pitch*

*a. Rate pitch.* The perceived pitch of stimulation increases with stimulation rate (the rate of pulses delivered to the implanted electrode or electrodes) from about 60 Hz through 350–400 Hz for most subjects. Below 60 Hz, subjects may report pulsating sounds (e.g., telephone ringing, ratchets and such). Above 400 Hz, there is fusion of the input pulses, with no further increase in pitch (Simmons, 1966; Bilger, 1977, 1983; Tong *et al.*, 1982; Townshend *et al.*, 1987). These sounds above fusion frequency often generate percepts of considerably higher pitch than the sensation a normal hearer might describe for a 400-Hz sine wave, sometimes by several octaves. Some subjects are able to discriminate slightly higher rates, but claims of rate discrimination at higher frequencies (such as 1000 Hz) may be attributable to changes in stimulus intensity.

The best frequency difference limens (DLs) are about 5% for stimuli at 100 Hz, 5%–10% at 200 Hz, and 8%–15% at 300 Hz, when collected with a two-alternative, forced-choice paradigm (Simmons, 1966; Hochmair-Desoyer *et al.*, 1983). These compare to approximately 0.2% for unimpaired people listening to acoustic waveforms (Wier *et al.*, 1977). Many patients do considerably worse. Waveform—sine versus rectangular pulses—does not alter DLs substantially. Most of the data have been collected from subjects at a "comfortable loudness." The effects of stimulus duration have not been thoroughly studied. The DLs do worsen slowly when burst durations are decreased below 100 ms and are probably optimal at 200–300 ms. DLs not only differ among subjects, but can also differ among electrodes on the same subject. The reliability of many DL measurements may be marred by contamination with loudness changes.

*b. Place pitch.* Subjects implanted with more than one electrode report (with varying degrees of conviction) that stimulation of individual electrodes by pulses having no rate-pitch information (single pulses or pulse trains above the fusion frequency) produces different pitch sensations (Townshend *et al.*, 1987). The ordering of these characteristic pitches roughly corresponds with the electrode location within the scala tympani. That is, when subjects are asked to rank pitch—as a floor-to-ceiling height, by anecdotes, or by a sharpness-dullness continuum—the lowest pitch is likely to be obtained at the most apical electrode and the highest pitch at the most basal electrode (Tong *et al.*, 1982). The sensations accompanying the stimulation itself, whether or not rate is also varied, are not described as "pure," in the sense of listening to a sinusoid, but seem rich in harmonics. One subject was able to compare electrically stimulated pitch with acoustic matching in his opposite, normal-hearing ear, but no exact matches were possible (Eddington *et al.*, 1978).

There are very few data available on the consistency of pitch ranking among subjects. It is clear that some perceive these electrode-specific sensations more easily than others. Some can "correctly" rank all electrodes and are seldom confused during paired-comparison trials. Others have electrodes in which the low-to-high ranking, apex to base, has discontinuities or local reversals. Still others report very little pitch difference among electrodes, and this is confirmed by poor or inconsistent ranking scores.

*c. Range of percepts.* The variety of anecdotal descriptions suggests ranges of several kilohertz for some subjects to only assorted "buzzes" for others. Both rate and place of stimulation affect anecdotal ranking decisions within the rate-pitch range (Eddington *et al.*, 1978; Atlas *et al.*, 1983). For example, a 200-Hz stimulus on one electrode may be ranked lower in pitch than 200 Hz on another electrode, and the corresponding anecdotal description might describe a large truck horn versus a car horn. There have been anecdotal reports of experimenters feeding back these presumed pitches as simple tunes, with limited success.

*d. Intensity effects.* In most, but not all subjects, stimulus intensity has important effects on pitch. The pitch can either increase or decrease with intensity, depending on the subject (Townshend *et al.*, 1987). Occasionally, pitch increments created by as little as a 2-dB intensity increment can produce anecdotal descriptions suggesting rises of an octave and more. These intensity-pitch effects, aside from not being totally predictable from one electrode to the next, have probably led to errors in some reported psychophysical results. It may be that pitch judgments are so level dependent that intensity must be randomized above and below the level of equally loud stimuli across frequency, to avoid confounding effects.

detection experiments, while others require very long silent intervals in excess of 100 ms (Hochmair-Desoyer *et al.*, 1983; Dobie and Dillier, 1985; Moore and Glasberg, 1988; Shannon, 1989).

This has led some researchers to speculate that people with very good temporal discrimination may be better candidates for single-channel implants than those with poor time discrimination. The use of this measurement as a preimplant diagnostic requires validation, however. It also would require some residual hearing, or a preliminary electrode placement.

#### 3. Percepts with multiple electrodes

*a. Electrode interactions.* Longitudinal spread of electrical charge among several electrodes stimulated simultaneously in the scala tympani is a significant but poorly resolved aspect of multielectrode stimulation. In general, bipolar stimulation between two closely spaced electrodes produces less interference than does monopolar stimulation. The principal index of such interactions has been loudness interactions, e.g., when two closely spaced electrodes, say 5 mm, are stimulated at their respective individual thresholds simultaneously, the resulting loudness is likely to be very much greater than one might expect from comparable acoustic loudness summation in a normal ear. Furthermore, such loudness changes are not always logically predictable in the same patient among different electrodes, or between patients (Shannon, 1983b).

At this writing, the adjustment of loudness for simultaneous multielectrode stimulation remains empirical. Some stimulation schemes purposefully avoid interaction by sequential, rather than simultaneous, stimulation.

*b. Other percepts with multiple-electrode stimulation.* Studies of speech or speech-sound discrimination have preempted studies of conventional psychoacoustics of simultaneous stimulation, except for threshold and loudness. It is not even entirely clear whether, when two electrodes are stimulated simultaneously, persons hear two separate sounds or a single fused percept.

### B. Risks involved in cochlear implants

The risks of a cochlear implant can be divided into the risks associated with any similar ear surgery performed under a general anesthetic, and those risks specific to the implant. For the surgery itself, there is a 1:5000 incidence of death or serious morbidity from a general anesthetic in an otherwise healthy individual. The possibility also exists of temporary or permanent injury to the auditory or vestibular nerve, of wound infection, and, in some implant procedures, disruption of the ossicular chain. Incidence data are not available on these risks; however none of these can be considered major hazards.

Specific to implants are the following:

(1) Damage to the inner ear resulting in loss of any residual hearing (this has occurred).

(2) Degeneration of ganglion cells to the point that the implant becomes nonfunctional. [While the incidence is unknown, there appears to be no evidence that short-term use (3 years or less) has caused degeneration to a nonfunctional state. Experience for longer times is confined to smaller numbers of patients, but there is also no evidence of degeneration after 5–6 years of use.]

(3) Temporary or permanent balance disturbances. (This has occurred in several patients temporarily after surgery, and there are patients who have experienced balance disturbances upon stimulation. Insofar as we are aware, none of these has been disabling.)

(4) Failure of electrical stimulation to produce sound. (This occurs.)

(5) Extrusion of the implant. (This has occurred.)

(6) Foreign body reaction or wound infections, including meningitis. (This occurs.)

(7) Device failure for a variety of reasons. (This occurs.)

(8) Inappropriate expectations by the implant patient regarding hearing results. (This occurs.)

(9) Possibility of a middle-ear infection entering the cochlea via the implant, then causing damage to the membranous labyrinth and/or meningitis. (To our knowledge, this has not occurred).

The discussion of negative results from cochlear implants should be more open. In fact, there is very little in the medical literature, other than some individual case reports and some data on animal models (e.g., Burgio, 1986; Leake-Jones and Rebscher, 1983; Sutton, 1984; Zrunek and Burian, 1985). Thus it is impossible to provide comprehensive incidence figures for associated risks. It is reasonable to presume that some of these risks increase with significant invasions of the cochlea, i.e., an electrode(s) extending more than about 6 mm into the scala tympani, or with the use of a "hardwired" percutaneous connector. Some risks may be decreased by so-called extracochlear stimulators, wherein the stimulating electrode is placed outside the cochlea, on the round window or round window niche. (Extracochlear devices may *increase* the likelihood of facial-nerve stimulation.) Whether or not these stimulators are as effective a communication aid as intracochlear devices is questioned by many practitioners. Too few patients have had extensive experience with the extracochlear devices to provide a solid, data-based evaluation of their performance (also see National Institutes of Health, 1988).

The above risks, on balance, do not appear to be serious in totally deafened adults. The question of long-term nerve degeneration associated with the use of intracochlear electrodes remains to be answered.

### C. Patient selection criteria

Originally, implantation was restricted by general agreement to postlingually deafened adults who could not benefit from a hearing aid. At this time, both the 3M House and Nucleus devices are approved for experimental trials for implantation in children of various ages. The House Ear Institute has also implanted prelingually deafened adults, the opposite ears of hearing-aid users, and at least a few patients with residual acoustic response in the implanted ear.

Virtually nothing is known at the present about the actual as opposed to the published criteria for implant candi-

TABLE IV. Perception of environmental sounds with cochlear implants (percentage correct responses) (SC = Single-channel; MC = Multiple-channel; IC = Intracochlear).

| Authors (date) | Device type | N | HRRC 20-item | MAC battery tests: Environment sounds | MAC battery tests: 20-item closed | MAC battery tests: 20-item open |
|---|---|---|---|---|---|---|
| Eisenberg *et al.* (1983) | SC–IC | 86 | 55.1 | ··· | ··· | ··· |
| Edgerton *et al.* (1983) | SC–IC | 10 | ··· | 40 | ··· | ··· |
| Tyler *et al.* (1985) | SC–IC | 3 | ··· | ··· | 60 | 28.3 |
| | SC–IC | 3 | ··· | ··· | 63.3 | 28.3 |
| Mecklenburg and Brimacombe (1985) | MC–IC | 37 | ··· | 31 | ··· | ··· |
| Tyler *et al.* (1984b) | MC–IC | 2 | ··· | ··· | 80 | 47.5 |
| Dowell *et al.* (1985a) | MC–IC | 6 | ··· | 27(13)[a] | ··· | ··· |
| Schindler and Kessler (1987) | MC–IC | 8 | ··· | 42 | ··· | ··· |
| Chance | | | ··· | ··· | 5 | ··· |

[a] "Modified" MAC test.

TABLE V. Speech discriminations with cochlear implants (percentage correct responses) (SC = single-channel; MC = Multiple-channel; IC = Intracochlear).

| Authors (date) | Device type | N | MAC noise/voice | Male/ female | Spondee same/different |
|---|---|---|---|---|---|
| Edgerton *et al.* (1983) | SC–IC | 11 | 70 | ··· | ··· |
| Tyler *et al.* (1985) | SC–IC | 3 | 58.7[a] | 71.7 | 75 |
| | SC–IC | 3 | 80[a] | 83.3 | 80 |
| Mecklenburg and | MC–IC | 33 | ··· | 85(66) | ··· |
| Brimacombe (1985) | | 38 | ··· | ··· | 85(63) |
| Tyler *et al.* (1984b) | MC–IC | 2 | 95[a] | 95 | ··· |
| Dowell *et al.* (1985a) | MC–IC | 6 | 96(69)[b] | 83(68) | 88(70) |
| Schindler and Kessler (1987) | MC–IC | 8 | 96 | ··· | 93 |
| Chance | | | 50 | 50 | 50 |

[a] Augmented test 40 items.
[b] "Modified" MAC tests.

TABLE VI. Perception of prosody cues with cochlear implants (percentage correct responses) (SC = Single-channel; MC = Multiple-channel; IC = Intracochlear).

| Authors (Date) | Device type | N | MTS test: Word | MTS test: Stress | MAC battery tests: Question statement | MAC battery tests: Accent | MAC battery tests: No. of syllables |
|---|---|---|---|---|---|---|---|
| Eisenberg *et al.* (1983) | SC–IC | 86 | 34.5 | 73.9 | ··· | ··· | ··· |
| Edgerton *et al.* (1983) | SC–IC | 12 | ··· | ··· | 65 | ··· | ··· |
| | | 11 | ··· | ··· | ··· | 55 | ··· |
| Tyler *et al.* (1985) | SC–IC | 3 | 16.3 | 75 | 58.3 | 43.3 | 57 |
| | | 3 | 29.3 | 60 | 68.3 | 36.7 | 66.7 |
| Mecklenburg and | MC–IC | 37 | ··· | ··· | 68(51) | ··· | ··· |
| Brimacombe (1985) | | 36 | ··· | ··· | ··· | 36(33) | ··· |
| Tyler *et al.* (1984b) | MC–IC | 2 | ··· | ··· | 57.5 | 50 | 56.4 |
| Eddington and Orth (1985) | MC–IC | 2 | ··· | ··· | 85 | ··· | ··· |
| Dowell *et al.* (1985a) | MC–IC | 6 | ··· | ··· | 46(49)[a] | 80(47) | 96(77) |
| Schindler and Kessler (1987) | MC–IC | 8 | ··· | ··· | 80 | 70 | ··· |
| Chance | | | 8.3 | 33 | 50 | 25 | 50 |

[a] "Modified" MAC tests.

enhancement depends, in part, on their baseline lipreading skill, which varies enormously across individuals. While some of the most successful individuals can recognize some words in sentences without visual cues, that remains the exception rather than the rule. The following discussion is therefore limited to the results of auditory-plus-visual testing.

The recognition of words in sentences is typically better than the recognition of isolated words, presumably due to the addition of grammatical and contextual cues. Several groups, particularly those groups who reported patients recognizing words in isolation, have noted a few subjects who could recognize more than 25% of the words in sentences (e.g., Dowell *et al.*, 1985a). The obtained values depend on the number of presentations of each test sentence, the internal predictability of the words in the sentences, and the manner of scoring (key words, every word, meaning).

Dowell *et al.* (1985a) reported that one patient out of six scored 38% on the CID everyday sentence test. The other patients got only two or three words correct. Tyler *et al.* (1984b) tested two additional patients with the Melbourne device; one obtained 33 percent correct recognition of words in sentences with a familiar speaker, and both scored about 40 percent correct when a picture of an object mentioned in the sentence was presented as a cue. Burian (1984) noted that 5 out of 14 postlingually deafened patients scored higher than 25 percent correct on sentences. Hochmair-Desoyer *et al.* (1985), reporting on what are presumably at least some of the same patients, reported that 7 out of 12 patients scored above 25 percent correct recognition of words in sentences.

### *7. Tracking*

De Filippo and Scott (1978) described a measure of proficiency in the reception of connected discourse, which has come to be known as tracking. In this method, the speaker (tester) reads aloud from a printed text, and the listener (subject) must repeat the message (a word, phrase, or sentence, depending on the level of training) verbatim. According to the original description of the method, errors must be resolved using a hierarchy of strategies (repeat the word or phrase, paraphrase or use synonyms, spell it, etc.), but only the verbal channel is to be used, with no recourse to purely visual input (charades, signing, finger-spelling, etc.). The speaker proceeds when all previous words have been correctly repeated. The subject's score is reported in terms of words per minute (wpm) correctly communicated. Normal subjects can perform at about 110 wpm (i.e., half the normal reading-aloud rate, since both tester and subject have to repeat all words); 70 wpm has been proposed as an approximate threshold of social adequacy (Levitt, 1985).

Perhaps the greatest advantages of this test are its simplicity and face validity. Only a trained tester, some reading material, and a stopwatch are required. Not only can this test be applied by all groups doing cochlear implant work, but it is also applicable to other sensory aid or substitution systems, such as tactile aids. Tracking has unusual face validity as a practical measure of communicative benefit, since it uses connected discourse. Compared with element-processing tests like those of word and phoneme recognition (Dent *et al.*, 1987), the tracking task is easier because of contextual cues but may be more difficult because of the requirement to process at high speeds (up to 10 phonemes/s in normal speech). A final advantage of the tracking method is that it can also be used as a form of therapy—both in rehabilitation classes and as homework.

However, there are certain difficulties in using the tracking method to analyze and compare results across groups (Tye-Murray and Tyler, 1988). Materials have not been standardized and obviously will vary in difficulty and resulting wpm scores. Because the primary goal is to assess the contribution of the implant as an aid to speechreading, the comparison between scores obtained by vision only and those obtained by vision plus stimulation is of greater importance than the absolute levels of either score. There is considerable variability in tracking scores for individual 5–10 min sessions, even after accounting for obvious sources of variability, such as degree of learning, different speakers, and different materials. Some of this variance appears to result from abrupt changes in the strategies used by speaker and the receiver. There appears to be no well-accepted convention for dealing with these problems, but most authors (admirably) present tracking results on a trial-by-trial basis.

Table VII summarizes the speech-tracking data obtained by several cochlear-implant groups. Where a certain group has published more than one paper including tracking data, the most recent has been chosen. A disappointing omission from the table is the Vienna group, which has reported no tracking data so far. Groups with all types of devices—single- or multichannel, intra or extracochlear—found that their systems provided substantial aid to speechreading for most or all of their subjects. When this was tested statistically (e.g., Robbins *et al.*, 1985; Dent *et al.*, 1985), the gains were clearly significant (i.e., large enough that they were probably not due to chance alone). There is considerable variability across subjects; one group (Robbins *et al.*, 1985) found that prior success with a hearing aid was correlated with ability to benefit from the implant. Inspection of the data suggests that some of the variability might also be reduced by stratifying subjects according to vision-only (VO scores) speechreading skills: the largest percentage gains are made by subjects who are poor speechreaders, while the largest gains in wpm seem to accrue to subjects who are fair-to-good speechreaders (VO = 10–30 wpm). Some subjects even show a decrement in VO performance over time, as they come to rely less on speechreading. (Other explanations are possible: for example, both the tester and the patient may become frustrated with the VO condition and may not try as hard as before). Clearly, it could be misleading to present only a sound-plus-vision to vision-only ratio $[(S+V)/VO]$ at asymptote as a measure of benefit, when this ratio can be artificially inflated by a reduction in VO performance. None of the subjects reported to date has achieved the 70 wpm level, but this must be tempered by the recognition that tracking rates for normal subjects (110 wpm) do not apply fairly to the testing of deaf subjects, who must look up from the written material, before speaking, to permit speechreading. Under these conditions, even normal subjects track at only about 75 wpm (Robbins *et al.*, 1985).

lation in the same subject with opposing results. However, in most studies there is a serious problem in experimental design: a patient with considerable experience with one stimulus mode (or "code") is unlikely to perform optimally during a relatively brief exposure to another. Until this issue is addressed, conflicting results will probably continue to be reported.

### H. Patient results and performance: Children

There is no issue in cochlear implants more controversial than the implantation of children (Simmons, 1985; Tyler *et al.*, 1987; Popelka and Gittelman, 1984). Many of the points of disagreement can only be alluded to here, but they include: the possibility of implanting a very young child who would have performed better with a hearing aid in the implanted ear than with the implant; unknown effects of long-term (lifelong) electrical stimulation; possible unreliability of implanted devices and the need for repeated surgery; uncertainty regarding the proper educational environment for implanted children (manual versus oral, or total-communication programs); and the chances of damaging a cochlea that might have been implanted with a greatly improved device at some future time.

Though cochlear implants have already gained acceptance for selected adult patients, there is a consensus that implantation criteria should be more conservative for children (see National Institutes of Health, 1988). Put another way, many clinicians and scientists agree that a higher benefit-risk ratio is required to implant children. In opposition to arguments against implanting children is the demonstrated importance of early experience on the development of language. It is possible that early electrical stimulation of the auditory system may be an important aid to auditory system development, to the acquisition of linguistic skills, or both.

Questions such as these need to be addressed by studies of the effects of implants in those children who do receive them. The difficulty with such studies is that they do not produce clear or quick answers because of: the difficulty of testing children's performance; the necessary longitudinal nature of the studies; the diversity of educational settings of the children; the difficulty of long-term follow-up; and the small number and relative heterogeneity of the samples of children available for this research.

Assessment of the benefits in those children who have been implanted has been hampered by two additional problems. One is that changes in speech-production skills, language, and educational achievement, while being desirable benefits of the implant, are influenced by many variables and this complicates the attribution of changes to the implant (Tyler *et al.*, 1987). Another problem relates to the tests that have been chosen by implant researchers and clinicians to measure benefit. Unfortunately, the validity of two of the most commonly used tests, the Discrimination After Training test, and the Test of Auditory Comprehension, has been questioned. The former may encourage the teaching of test items, and the latter is heavily dependent on cognitive maturation.

In this section, only a small amount of data has been represented, since there are few published studies. More studies are now being conducted and should be published within the next few years. The categories of benefit discussed here relate best to adult users of cochlear implants. Anecdotal reports of diminished hyperactivity and similar positive effects of the implant have not yet been quantified and are not considered here. Although many researchers and parents have high hopes for the effectiveness of cochlear implants in children, the absence of appropriate research makes it as yet impossible to know whether these expectations are valid. There are currently two investigational devices being implanted in children in the United States: the 3M/House and the Nucleus devices (Berliner and Eisenberg, 1987; Clark *et al.*, 1987). Implantation of children with the 3M/House device began in 1980, and there are now over 270 children that have been implanted. (Manufacture of the 3M/House device had been discontinued at the time of printing of this article.) The Nucleus children's implant program began in 1987. Early tests of children's abilities with implants did not show maximum scores on elemental closed-set tests such as the MTS test. More recently, however, open-set discrimination testing (on a highly selected group of children) has been reported in which 9 of 10 children with the 3M/House implant showed some open-set word recognition and 10 of 19 showed open-set comprehension on items from the Glendonald auditory screening procedure, administered without visual cues (Berliner and Eisenberg, 1987). Recent testing by Moog and Geers (1988) has confirmed, though primarily on closed-set tests, that at least some children with cochlear implants perform remarkably well. The natural development of children's sound and language capabilities means that assessing the value of the implant to deaf children requires long-term study. The results reported here represent only the beginning of those that should be obtained.

#### *1. Environmental sounds*

Thielemeir *et al.* (1985) indicated that the average child ($N = 32$) tested 1-year following implant could perform at level 1 of the test of auditory comprehension. At this level of the test, the subject must discriminate between linguistic versus human nonlinguistic versus environmental sounds. This does not imply that the children could recognize environmental sounds, since gross time-intensity cues could be used to distinguish speech from nonspeech stimuli. Although the specific number is not stated, some children could not perform this task following implant. Popelka and Gittelman (1984) reported data for one 8-year-old boy who had received the 3M/House implant and scored at chance on this task. Berliner and Eisenberg (1987) reported that one of their better implanted postlinguistic children (age 15 years) was able to score 95 percent correct on the House Ear Institute environmental sound test.

#### *2. Stress/prosody*

Thielemeir *et al.* (1985) reported that the average score of 30 children tested at 1 year following implant was at level 6.8 on the Discrimination After Training test (Thielemeir, 1984). The levels of this test progress on a hierarchy of primarily prosodic discriminations. The mean level for 106 children before implant was 2.2. This suggests that the children may be making progress in using the stimulation pro-

In addition, it is normally more heavily used than the tactual sense. Thus it is not clear *a priori* which of the two senses is most appropriate for exploitation as a hearing substitute. For those individuals who are blind as well as deaf, of course, there is no choice: only the tactual sense is available (since smell and taste are clearly inadequate because of their inability to follow rapidly changing stimuli).

In general, the communication methods that employ nonauditory displays can be subdivided into "synthetic" and "natural." The synthetic methods make use of acoustic inputs and require a device to transform the acoustic energy into a visual or tactual stimulus. These methods are products of the scientific research community and, for the most part, are still in the experimental stage. The natural methods make use of nonacoustic inputs and do not require any transformation device. Most of these methods were born of necessity within the deaf community and have been used extensively for many years. Whereas the synthetic methods operate on acoustic signals, the natural methods operate on nonacoustic signals associated with the acoustic signals (e.g., the lipreading signals associated with talking) or nonacoustic signals that are specially designed for deaf individuals (e.g., the signs in sign language). The former type of natural method, e.g., lipreading, will be referred to as *natural-general* and the latter type (e.g., signing), which requires special knowledge or behavior on the part of the sender, as *natural-special*.

The synthetic methods have greater ultimate potential as substitutes for hearing because they, like hearing, directly sense the acoustical environment. The natural methods are important, however, because they are the methods actually being used by the deaf population (and the methods in which subjects have received long-term training) and because they provide important information on the capacity of the visual and tactual senses as substitutes for hearing. In addition, one of these methods, namely visual speechreading (lipreading), is usually assumed to be available to the user when considering any type of speech-perception aid. Independent of whether the aid is a tactile aid, a conventional hearing aid, or a cochlear-implant aid, when the hearing loss is profound the aid is usually envisioned as an aid to lipreading.

The relative attractiveness of the visual versus the tactual channel is different for natural methods and synthetic methods. Whereas in the natural domain vision has the advantages of not requiring direct physical contact and of functioning at a distance, in the synthetic domain (where the input is acoustic for both visual and tactual systems), this advantage disappears. This distinction is reflected in the fact that natural visual methods are used by essentially all deaf individuals who are not blind, while research on synthetic methods has mostly included work on tactual displays.

It should also be noted that the research groups that have been involved in development of synthetic systems, i.e., sensory-substitution aids, differ in certain ways from those involved in the development of cochlear implants. Whereas the latter effort has been largely an effort of the medical/clinical community and has received extensive financial and industrial support, the former is primarily the work of university-based investigators and has the characteristics, typically, of more academically oriented research. One of these academic characteristics is unwillingness to attempt full-scale clinical trials on impaired individuals until very late in the development of new devices.

### B. Postlingual versus prelingual deafness

The relative usefulness of sensory-substitution aids versus cochlear implants may depend to a great extent on whether the patient is postlingually or prelingually deaf. Because of the great importance of this issue, it is discussed before reviewing the properties of sensory substitution aids.

#### *1. Postlingually deaf people*

For patients who previously had a sense of hearing and learned language before the onset of deafness (postlingual deafness), the cochlear implant has the advantage of providing the patient with the important psychological benefit of "hearing again" (i.e., of experiencing some form of auditory sensation). Thus, even if a sensory-substitution aid resulted in equivalent communication performance, it might not provide equivalent subjective satisfaction. Furthermore, for postlingually deafened patients, the amount of learning/training required to fully exploit a sensory-substitution aid is likely to be greater than that required to fully exploit a cochlear implant. With the sensory-substitution aid, one must learn not only to distinguish among the sensations produced by the aid, but also to correctly match these sensations with external acoustic events. Although there is undoubtedly some transference across senses, and some learning of this sort must also take place with the implant, for this class of patients the amount of learning that is required for optimal performance is likely to be less with the implant. To date, most patients treated with cochlear implants fall into the postlingual category.

#### *2. People deaf since birth*

For patients who have been deaf since birth, the above advantages of cochlear implants may be diminished. A patient to whom a sensory-substitution aid is applied at birth, like a patient to whom a cochlear implant is applied at birth may, develop a sense of hearing that meets scientifically meaningful criteria for that sensory modality. Not only will both patients "hear" according to an objective, behavioral definition of hearing (with a discriminative capacity that depends on the resolving power of the prosthetic device and the stimulated sensory apparatus), but both patients will experience the impinging acoustic energy in terms of externalized object/events rather than feelings in or on their own bodies. Just as a normal-hearing person does not localize an acoustic stimulus in his or her ear (and becomes conscious of the role played by the ear in sensing the acoustic environment only when the relation between the ear and this environment is altered), it seems very likely that the person who has grown up with a tactile aid probably will not localize an acoustic stimulus on the body surface stimulated by the tactile aid. It is also possible that the advantage of cochlear implants with respect to the above-mentioned learning problem might disappear. For both the sensory-substitution aid and the cochlear implant, the learning required to construct

that occur in speech reception via Tadoma can be explained by its failure to make available information about tongue position.

Results on normal subjects with simulated deafness and blindness (Reed *et al.*, 1978, 1982a,b) indicate that the deaf-blind subjects have no special tactile sensitivity: performance of the two groups is roughly equivalent on tests of basic tactile discrimination abilities. These results also suggest that learning Tadoma (for someone who already has language knowledge) is roughly comparable in difficulty to learning a complex foreign language.

In the same manner that lipreading demonstrates the adequacy of the visual system for receiving spoken speech, Tadoma demonstrates the adequacy of the tactual system for receiving spoken speech. As will be seen in subsequent sections, the performance achieved with these methods provides merely a lower bound; performance can be improved (in the visual case, and probably also in the tactual case) by augmenting the speechreading stimulus with other stimuli.

*c. Visual and tactual combined.* At least one individual, who is deaf but not blind, achieves excellent results by supplementing visual speechreading information with tactual information obtained by placing a hand on the shoulder and neck of the talker (Plant and Spens, 1986). The improvements obtained with this method, which the subject has used for 40 years, over visual speech reading alone are 88% versus 44% for consonant identification, 66% versus 33% for open-set word identification, 98% versus 85% for scores in CID sentences, and 63 versus 42 wpm in continuous discourse tracking. The results obtained when the hand is removed from the shoulder and stimulated instead with an experimental artificial tactile aid are intermediate between these two extremes. However, the total exposure time for each of the two aids tested was less than 4 h.

#### 2. Cued speech

In cued speech, hand configurations are used to disambiguate stimuli that are confused in speechreading (Cornett, 1967). Eight hand configurations are used to code consonants and four locations to code vowels, and the hand cues are presented in synchrony with the spoken speech. Cued speech was designed to supplement visual speechreading and to date has been used only in the visual sense. The extent to which these same hand configurations can be used to improve tactual speechreading is only now being studied (Reed *et al.*, 1987b).

The effectiveness of cued speech as a supplement to lipreading is illustrated by the study of Nicholls and Ling (1982) on 18 children with profound losses who were trained in this method for at least 4 years. On the average, identification performance with nonsense syllables improved from 30 or 36 percent correct with lipreading or cueing alone to 84 percent correct with the two combined. Similarly, for identification of the final word in low-predictability sentences, the scores were approximately 25% (lipreading alone), 43% (cueing alone), and 97% (lipreading plus cueing). As stated by Nicholls and Ling (1982:267–268):

> The subjects' responses under the combined condition were outstandingly and uniformly good and merit considerable attention.... The system enabled all of the subjects to receive precise phonemic and linguistic information both at a syllabic level and in running speech. Speech reception at an equally high level of accuracy by profoundly and totally hearing-impaired children has not been previously reported. The children's average scores ... are within the range of normal-hearing listeners' reception of similar materials through audition.

#### 3. Fingerspelling

Fingerspelling is a system in which words are spelled out letter-by-letter using handshapes that correspond to the letters. It is sometimes used as a complete communication system by itself and sometimes as a supplement to American Sign Language (e.g., to communicate proper nouns for which signs have not been developed). Unlike cued speech but like signing, fingerspelling is used in the tactual mode as well as the visual mode.

At a typical transmission rate of five letters/s (roughly one-third the normal speaking rate), trained subjects can receive fingerspelling visually with negligible error rates. For example, for open-set identification of words in isolation, scores usually fall in the range 90–100 percent correct (e.g., Zakia and Haber, 1971; Thomson, 1984). At a rate of 15 letters/s, however, scores are reduced below 50 percent correct (Thomson, 1984). Roughly speaking, the function describing the dependence of percentage-correct on rate of presentation for visual fingerspelling appears to be similar to the equivalent function for windowed reading (reading through a hole or window in a mask that is moved across the text) with a window width of two–four letters (Thomson, 1984). The errors made in visual fingerspelling generally occur among letters represented by similar handshapes, finger direction, and finger identity (e.g., see Hawes and Danhauer, 1980).

In a recently initiated study of the tactual reception of fingerspelling (Reed *et al.*, 1986, 1987a), five deaf-blind subjects achieved perfect scores on conversational sentences presented at rates of five letters/s and were able to achieve scores on the tracking test (De Filippo and Scott, 1978) of roughly 30 words/min (a result similar to that obtained by experienced Tadoma users).

#### 4. Signing

The major varieties of sign language (used by roughly half a million people in the United States) are often described as falling on a scale that extends from manually coded or signed English (SE) to Pidgin Sign English (PSE) to American Sign Language (ASL). ASL is a natural and complete sign language; it not only has a grammar and vocabulary that is entirely distinct from English, but it employs communication techniques that differ from those used in any spoken language. SE makes use of manual signs to represent the English language; such systems are often devised

Variations of the straightforward spectrographic approach, directly concerned with impaired listeners but oriented toward speech-production training rather than speech reception, have also been studied (e.g., see the reviews by Pickett, 1968, 1969; Levitt, 1973, 1985). In accordance with the speech-training goal, these systems tend to incorporate processing algorithms and displays that permit the observer to focus on selected characteristics of the speech signal, and they do not operate in real time.

In a project designed to explore the use of speech reception of articulatory features complementary to those obtainable through residual low-frequency hearing, a real-time visual display of place and manner information (hand extracted from the speech waveforms) was tested with and without low-frequency auditory information (Goldberg, 1972). Reception of speech segments for both normal and hearing-impaired listeners was found to improve with the visual display (e.g., recognition of 48 consonant-vowel-consonant [CVC] stimuli changed from 40% to 75% when the visual display was added). Significant transfer of learning (one of the main goals of the study) was also observed. How the results would have been affected if lipreading had been included is unknown.

Two speech-feature-extraction systems that were designed for practical use in speech reception are the Upton Eyeglass and the Autocuer. The *Upton Eyeglass* (Upton, 1968) contains a signal processing scheme (analyzer) and a set of miniature lights (display) that are designed to convey to the user characteristics of the incoming sound, such as voicing or frication, that are difficult to lipread. The analyzer consists of filters and logic circuits that classify the sound, and the display consists of a light-emitting diode array in the form of a block figure 8 that is superimposed (by the use of a mirror) on the face of the talker. Although little systematic evaluation of this device has been completed, some results are available (Pickett *et al.*, 1974; Gengel, 1976, 1982). An improvement of roughly 16% over lipreading alone was exhibited for monosyllabic words by the poorer lipreaders in a group of hearing-impaired college students who were trained and tested with the aid for six hours. However, the better lipreaders showed less improvement on these words, and none of the subjects showed significant improvement for continuous discourse. In one case study, after roughly 6 months of aid use, key word scores for unfamiliar sentences showed an improvement of approximately 18% when the Upton aid was combined with lipreading or with amplification, and 13% when it was combined with both lipreading and amplification (Gengel, 1982). These improvements are similar to those achieved by Upton himself with the same display, but smaller than those achieved (25%–30%) with a modified display that employs color as well as spatial coding.

The *Autocuer* (Cornett *et al.*, 1982) is similar in that it too makes use of LEDs (miniature lights) and an optical system mounted on eyeglasses to project near the mouth of the talker, an image that displays the results of a classificatory analysis of the incoming speech sound. Two important differences, however, are the modern technology incorporated into the autocuer (i.e., microcomputer, integrated circuits) and the use of cued speech (which is known to be successful) as a starting point for the system design. Unfortunately, no adequate evaluation of the system has yet been performed. Pilot studies indicate that significant improvements in lipreading may be possible; however, the reliability of the automatic speech-feature-extraction system, the ability to understand continuous discourse at normal rates with the display even without any processing errors, and the influence of processing errors on speech-reception performance all await careful quantitative determination.

### 2. Tactual

General reviews of research on the use of the tactual sense as a substitute for hearing are available in Kirman (1973), Reed *et al.* (1982a), and Sherrick (1984). In essentially all cases, the tactual displays are designed to be applied to the skin and to stimulate cutaneous receptors. (Throughout the rest of this discussion, we refer to such displays as *tactile*). The possibility of developing synthetic tactual systems that make use of kinesthetic receptors (those that provide information about relative movement of parts of the body) for communication purposes is only now being explored (Loomis and Lederman, 1986).

The tactile displays are usually based on homogeneous arrays of vibrators or electrocutaneous stimulators. The array may consist of only a single element or include scores of elements arranged in a rectangular matrix. Body sites to which tactile displays have been applied include the fingertip, hand, wrist, forearm, collarbone, thigh, stomach, and recently the pinna. Different body sites have different spatial resolution and different-sized cortical representations. The extent to which different sites can be equated by appropriate scaling of element size and interelement distances in the array is not yet clear. For example, the effect of body site on temporal resolution (of great importance for speech reception) has not yet been determined.

Also poorly understood but potentially of great importance is the developmental plasticity of the tactile portions of the central nervous system. It may be important to provide the substitute device or aid early in life in order that the special neural circuitry required for optimal use of the device can develop.

As with the visual sense, substantial tactile research has been conducted on spectral displays (Reed *et al.*, 1982a). Since the tactile sense is relatively insensitive to the frequency composition of the stimulating waveform, spectral displays in the tactile sense use a frequency-to-place transformation: the outputs of the filters used to achieve the spectral decomposition are applied to different regions of skin. In a very rough sense, this frequency-to-place transformation is similar to that performed by the cochlea in the ear.

*a. Spectral displays.* Spectral displays of speech stimuli have been examined in a wide variety of studies [see the references cited in Reed *et al.* (1982a); the experimental survey by Spens (1980); and the work by Greene *et al.* (1983); Brooks and Frost (1983); Spens (1984); Brooks (1984); Brooks *et al.* (1985); Craig *et al.* (1985); Blamey and Clark (1985); Brooks *et al.* (1986a,b); Potts and Weisenberger (1987); and Weisenberger (1987)]. Aside from variations in the body site and in the type of stimulation (mechanical

1983; Spens and Plant, 1983; Proctor and Goldstein, 1983; Goldstein and Proctor, 1985; Geers, 1986). Of particular interest are reports concerning the effects of single-channel aids on the development of speech and language in prelingually deaf children. Proctor and Goldstein (1983) and Geers (1986) each introduced such an aid to a prelingually deaf child (approximately 3 years of age) and reported increased spontaneous vocalization and rapid growth of receptive vocabulary. Results on two slightly older children (Goldstein and Proctor, 1985) indicated significant improvements on a test of auditory language comprehension (Carrow, 1973) after 11 months of experience with the aid. Friel-Patti and Roeser (1983) evaluated the effects of a single-channel aid on the communication skills of four prelingually deaf children whose average age was 4 years. The amount of vocalization plus signing initiated by the children increased over the period and the aid was worn and appeared to decrease following its removal. In general, these results are consistent with the earlier study of Goldstein and Stark (1976), who demonstrated (using a vibrotactile array) an increase in the production of consonant-vowel utterances by prelingually deaf children aged 2 to 4 years who received speech-production training with the tactile aid compared with a control group who received identical training without the aid.

*c. Single-channel vibrotactile systems compared with cochlear implants.* Further data on single-channel systems, primarily for postlingually deaf adults, is becoming available in connection with cochlear-implant evaluations. Increasingly, investigators are attempting to compare the reception performance achieved with the implant to performance achieved with a tactile aid (e.g., Carney, 1984; Tyler *et al.*, 1984b; Blamey *et al.*, 1985; Miyamoto *et al.*, 1987). In the study by Tyler *et al.*, which compared cochlear implants (both single and multichannel) with a single-channel vibratory aid applied to eight normal-hearing subjects, the best results achieved with the tactile aid were roughly equivalent to the best results achieved with any of the implants (averaged over the ten tests performed). Except for certain prosodic tests, however, the worst results obtained with the tactile aid were inferior to the worst obtained with the implants. When the comparison between vibrotactile aid and implants was made on the implanted patients (using a test of syllabic-stress patterns, a male versus female test, and a four-choice spondee test), the results with the tactile aid were inferior. However, whereas the subjects had at least 3 months experience with their implants, they had only 1 h of training with the single-channel tactile aid.

In the study by Blamey *et al.*, which compared the benefits of a hand-held bone vibrator with those of a conventional high-powered hearing aid in postlingually deaf adults who were prospective cochlear implant patients, the results showed roughly equivalent positive effects for closed-set speech tests without lipreading, but no positive effects for open-set tests or tests with lipreading. Subjects who subsequently received implants (multichannel) demonstrated significantly improved performance with the implant, including performance on open-set tests and tests with lipreading. In the study by Carney (1984), the results of comparing a single-channel implant with a single-channel tactile aid applied to normal-hearing subjects showed no dramatic differences, although other differences between the young normal subjects and the implant wearers may have influenced this result. In the study by Miyamoto *et al.* (1987), concerned with the use of tactile aids in the evaluation procedure for cochlear implant candidacy, a number of wearable tactile aids (one or two channels) were compared with implants (single channel). Although many tests produced roughly equivalent results for the two types of prosthetics, some showed distinctly superior results for the implants. The authors concluded their report by observing that "... the question is not which device, cochlear implant or tactile device is better, but which device is more appropriate for which patient." An attempt to summarize comparative information on implants and tactile aids (multichannel as well as single channel) has been made by Pickett and McFarland (1985). A discussion of alternatives to cochlear implants that includes comments on ethics, cost, and clinical suitability is available in Martin (1983).

In a thoughtful and clinically relevant overview of results obtained with deaf children, Moog and Geers (1986) examined the benefits of single-channel vibrotactile aids and cochlear implants relative to performance obtained with conventional hearing aids. Based on the study of ten children (ages 2–13) who showed no ability to discriminate sounds on the basis of spectral information and had very poor speech reception through their hearing aids, Moog and Geers concluded that the benefits of the two types of single-channel devices were roughly equivalent, that these benefits were most clearly evident in "getting young children started in learning spoken language" (see also Richardson, 1986), and that substantially improved results would be obtained with multichannel devices that provide spectral information.

*d. "Binaural" vibrotactile aid.* In another unusual study, Weisenberger *et al.* (1987) explored the performance of a "binaural" vibrotactile aid in which different vibratory stimuli were applied to the skin of each ear canal by vibrating earmolds. (There is evidence that responses by profoundly deaf individuals to stimuli produced by ordinary hearing aids at very high levels result from excitation of the tactile system—e.g., Nober, 1967; however, in the present study, the system was designed for tactile stimulation at all levels.) It was thought that the use of the same site for the tactile aid as for a conventional hearing aid might increase the acceptability of the tactile device as an aid to hearing and also facilitate the use of hybrid systems involving both auditory and tactile stimulation. Both earmolds in the binaural aid tested provided crude spectral information by vibrating at 80 Hz in response to low-frequency acoustic energy and at 300 Hz in response to high-frequency acoustic energy. Tests on a few normal-hearing and impaired listeners, concerned with sound source localization, environmental sound identification, and syllable rhythm and stress identification, showed mixed results. The performance for the latter two tasks was superior to that for the localization task. It is uncertain, however, whether performance was superior to that which would have been achieved with conventional single-channel

the medical community are not currently well-informed in this area. Termination of the initial trial period will occur when the implant decision is made or earlier, perhaps, if the patient judges the aid to be worthless. Ideally, the trial period should be continued as long as the increased exposure and associated opportunities to learn the aid are leading to improved performance. In practice, however, a period of 3–6 months may be more appropriate. During this period, tests can be performed with this aid that are identical to tests used for evaluation of cochlear implants so that direct comparisons can be made, and so that the patient can understand the significance of the cochlear-implant tests. In all cases, the trial period should include a training and testing program designed by an appropriately experienced therapist.

The ideal clinical path for younger children (age less than 12 years) could be similar to that described above for adults, but would require more parental input. Before this ideal path can be specified with confidence, however, additional data on the performance of both cochlear implants and sensory-substitution aids in these younger patients is required.

Commercial firms that are producing wearable tactile aids are listed in the Appendix.

## IV. CONCLUSIONS AND RECOMMENDATIONS

The questions the working group was asked to address were, most simply, "What speech-perception aid is best for whom, and what help can one expect from the most appropriate aid?" Individual chapters have treated in considerable detail each of the three major types of aids. This chapter focuses on the classes of impaired listeners for whom the various aids appear to be most appropriate, and on the kinds of help each class of listeners may expect. In a final section, the working group has identified a number of questions, both basic and applied, that need to be answered before more successful speech-perception aids can be developed.

### A. Who is a candidate for what aid?

#### *1. Conventional electroacoustic aids versus cochlear implants and sensory-substitution aids*

This question deserves a data-based answer, at least a statistical or probabilistic one. There are, however, few studies that have systematically tried more than one type of aid on individual patients. Differences in the ability to identify a talker's intended message are often great among individuals with audiometrically identical hearing losses. It is often difficult to determine whether good performance reflects a superior aid or a superior listener. The data that are available suggest that a significant amount of the variation in performance is a function of the individual rather than of the aid or of the care with which the aid is fitted or adjusted. Nevertheless, the single variable that most strongly determines a listener's ability to perceive speech accurately through an electroacoustic hearing aid or, correspondingly, that establishes the need to consider other types of aids (cochlear implants or sensory-substitution aids) is *the portion of the speech spectrum (at the output of the acoustic aid) that the listener can successfully process.* This variable, in turn, is determined primarily by the degree and type of hearing loss. (In this chapter all references to hearing loss refer to sensorineural loss. Conductive hearing loss can generally be treated successfully either through surgery, medication, or by sufficient acoustic amplification to compensate for the loss.)

The working group concludes that individuals whose speech-frequency thresholds show mild-to-severe hearing loss (25–90 dB) will receive more usable speech-waveform detail through a conventional acoustic aid than from either of the two classes of aid that transform sound into nonacoustic stimulation. We stress that this means that the appropriate aid for the overwhelming majority of hearing impaired persons is the conventional electroacoustic aid, and that will continue to be true unless very substantial improvements are made in the other types of aids.

For those with speech-frequency losses in excess of 115 dB, the working group concludes that such people will derive little help from acoustic amplification. (Note, however, that a conventional aid, if visible to the talker, may cause the talker to speak distinctly in full view of the impaired listener and thereby to facilitate lipreading. Such modifications in the behavior of the talker can provide substantial improvements in communication.) The person who receives no significant help from acoustic amplification is clearly a candidate either for a cochlear implant or for one of the sensory-substitution aids. By "no significant help" is meant that electroacoustic amplification fails even to usefully enhance the speechreading (lip reading) of normal conversation. Recommendations for individuals with losses in the region 90–115 dB, a region in which the listener may receive small but significant benefits from acoustic amplification, are, as one would expect, more controversial and are highly dependent on the needs and expectations of the individual impaired listener and on the person making the recommendation.

#### *2. Cochlear implants versus sensory-substitution aids*

As indicated above, there is general agreement about the limitations of conventional electroacoustic aids for cases of profound hearing loss. There is not such general agreement, however, on the relative merits of the different possible treatments for such cases. In addition to the fundamental choice between oral and manual communication (see the comment in the Introduction), there are many areas of disagreement concerning the relative merits of cochlear implants and sensory-substitution aids within the oral approach. Unfortunately, too few patients have been systematically trained and tested with more than one class of aid to make available the controlled data base required to address this question directly. Generally speaking, physicians and scientists participating in implant programs consider individuals with profound losses to be candidates for implants, barring some complicating physical or psychological condition. However, other scientists, particularly those whose experience includes research on sensory-substitution aids, are not convinced that the implant should be the *only* form of aid recommended to the profoundly deaf. As noted in the Introduction, some groups have considered superior performance after brief exposure with a vibrotactile aid an indicator that the patient is

tution aids as well as implants can lead to a true replacement for hearing (in the sense described near the end of Sec. III).

(d) The question of whether to use intracochlear implants in young children cannot be answered unequivocally. The most positive view is that this is the single treatment most likely to result in successful oral-aural speech production and perception. The most negative view is that implants are unjustified when there are safer, much less expensive alternatives available, and no clear evidence has yet proven these alternatives to be less effective aids to speech development. Working group members with clinical or research experience with cochlear implants generally support the FDA decision to permit small-scale studies of children implanted with intracochlear implants.

(e) To the extent that a given prelingually deafened adult can be regarded as having no significant previous auditory experience, the advantage of treatment by cochlear implant is greatly reduced. Not only may such an individual be totally uninterested in a speech-communication aid of any kind (because of an allegiance to the world of manual communication), but since no sense of hearing was ever present, the notion of restoring this form of sensory experience has no meaning. In other words, one of the main advantages of cochlear implants over sensory-substitution aids for postlingually deafened adults may not apply to prelingually deafened adults.

(f) Patients who are doing well with a tactile aid and continue to show improvements should probably not be implanted until a sufficient trial is completed. There is no clear reason to implant a patient who achieves successful communication with a tactile aid.

(g) Specific recommendations for elderly patients are contained in the recent report of the CHABA Working Group on Speech Understanding and the Aging (1988).

*d. The research-treatment context.* Although laboratory studies indicate overlap between performance with cochlear implants and performance with vibrotactile aids, and although vibrotactile aids cost much less (by roughly a factor of 10) and involve no surgery, they have not been commercially available until the last few years. Consequently, the pool of patients with substantial experience is much smaller for vibrotactile aids than for the cochlear implants. Numerous reasons have been proposed to explain the rapid growth of cochlear implants versus alternative devices. One explanation that is frequently proposed is that cochlear implants were developed within the medical community as a means of helping patients for whom no other currently available treatment was successful, whereas sensory-substitution aids have been mainly a product of academic research, the strongest orientation of which is toward the discovery of general scientific principles. Research scientists have neither the credentials required to test new devices on patients nor, apparently, very strong inclinations to collaborate with others in doing so before all possible shortcomings have been eliminated. Had this been the orientation of the medical community, it seems likely that no patients would yet have been implanted. Differences in the goals and strategies of the clinician versus those of the scientist are difficult to document in any objective manner. The working group is convinced, however, that those differences, possibly as strongly as differences in effectiveness, have contributed to the rapid increase in the number of patients implanted and the relatively sluggish progress of alternative solutions, especially vibrotactile and visual aids.

### B. Expectations and hopes for the future

#### *1. Improved devices*

The working group believes that, for each class of aids that has been considered, substantial improvements in speech-reception performance can be achieved by improving the design of the aids. Focusing first on conventional acoustic aids, for example, we note the intense current research and development activity in the area of adaptive background noise reduction. Based on preliminary results in this area, we believe it highly probable that hearing aids with significantly improved noise reduction will be commercially available in the near future. Furthermore, since these noise-reduction schemes are essentially independent of the form of stimulation, once it has been demonstrated that they are useful for hearing aids, they undoubtedly will be incorporated into cochlear implants and sensory-substitution aids as well.

Another major thrust, relevant to cochlear implants and sensory-substitution aids, concerns the development and evaluation of multichannel systems. Theoretical considerations, as well as preliminary experimental results, indicate that aid performance can be substantially improved by partitioning the frequency spectrum into a number of frequency bands and stimulating distinct regions (in the cochlea or on the skin) with the signals derived from the various frequency channels. It is unlikely, however, that the following prediction made by Parkins and Anderson in 1983 (a prediction stated for cochlear implants, but also relevant to sensory-substitution aids) will be completely satisfied unless a significant increment in research effort occurs in the next few years (Parkins and Anderson, 1983:530):

> We believe that it is possible to develop a cochlear prosthesis that will provide excellent speech discrimination for a significant number of implanted patients without use of visual cues and that such a device will be available in the next ten years.

In addition to multichannel systems based on frequency decomposition, multichannel systems based on other parameters are being considered for systems that stimulate the tactile/kinesthetic senses (inspired by the results on Tadoma, as described in Sec. III). In these systems, not only are the speech signals decomposed into distinct channels on bases other than frequency content, but the signals are also presented using displays that are perceptually richer than the homogeneous vibrator arrays normally used with the frequency-decomposition systems. Whether practical systems of this type can be developed for clinical application and the extent to which such practical systems actually lead to improved speech reception remains to be seen.

Research is also being accelerated on the use of automatic speech-recognition systems for speech-reception aids. If one regards the aid primarily as a supplement to lipread-

The third and final conclusion is that there are many reasons to have high hopes for the future. Current research results on multichannel systems and on noise-reduction schemes (as well as on automatic speech recognition), together with the demonstration that the tactile/kinesthetic senses are adequate for receiving speech provided the user is thoroughly trained at a sufficiently early age and the reports of star implant users, all suggest that major improvements are within our grasp. If the research effort on this problem continues at its present rate, the next decade will yield aids that are substantially more effective than those now available. This is an important conclusion, because for many hearing-impaired persons only a modest increment in the current effectiveness of aids would be required for them to become socially adequate communicators (in the aural–oral mode). Unlike Parkins and Anderson, however, we do not feel confident enough to predict a truly successful auditory prosthesis for the profoundly deaf in a decade.

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## APPENDIX: COMMERCIAL AVAILABILITY

This Appendix lists the major characteristics of five of the most widely used implant devices in the United States as well as commercial firms that are producing wearable tactile aids, in 1986–1987.

### 1. Implant devices

Symbion/Ineraid: Four-electrode analog bandpass filtered, simultaneous stimulation, monopolar intracochlear scala tympani (22 mm), percutaneous connection to electrodes.

Nucleus: 21-channel digital sequential stimulation with speech feature extraction coding (stimulation rate is determined by voicing rate). The peak energy is extracted in 10-ms epochs and used to determine which of the 21 electrode pairs will be stimulated (according to a predetermined "low to high" electrode, pitch ranking); a single electrode is stimulated at a time. Intracochlear scala tympani (25 mm), demodulated, implanted electronics.

3M House: Single channel amplitude-modulated 16-kHz carrier, passive implanted electronics without demodulation analog stimulation limited to 340–2700 Hz, intracochlear (6 mm). (No longer available in 1991, but service is provided.)

3M Vienna: Single electrode analog, with wideband equalization, implanted electronics with carrier demodulation, round window implant site.

UCSF/Storz: Four-electrode analog band-pass filtered, bipolar stimulation, scala tympani insertion (22–25 mm), active implanted electronics. (Note: in 1988 this device was not in production but may be manufactured by another company.)

Table AI provides summary information about implant devices; more detailed statements from manufacturers have been published in ASHA, 27: 27-34, 1985.

TABLE AI. Implant devices available in the United States.

| Manufacturer/ device | Number of channels | Mono/ bipolar | Analog pulse | Implant site | Insertion depth |
|---|---|---|---|---|---|
| Symbion/ Ineraid | 4 | mono | analog | intra | 22 mm |
| Nucleus | 21 | bi | pulse | intra | 25 mm |
| 3M House | 1 | mono | analog | intra | 6 mm |
| 3M Vienna | 1 | mono | analog | intra | 6 mm |
| | | extra | 0 mm | | |
| UCSF/Storz | 4 | bi | analog | intra | 22–25 mm |

### 2. Tactile aids

Commercial firms that are producing wearable, portable, tactile aids include: AB Special Instruments (Minivib 3), Stockholm, Sweden; Audiological Engineering (Tactaid), Cambridge, Mass.; Coulter Associates (Portapitch), Washington D.C.; Siemens Hearing Instruments (Mini-Fonator) Union, N.J.; Tacticon Corp. (Tacticon), San Rafael, Calif.; Telex Communications (Televibe), Minneapolis, Minn. We do not know of any wearable, portable, visual aid that is available commercially, although this situation may change in the not-too-distant future with commercial introduction of the Autocuer (R. Beadles, personal communication, 1986).

[1] Although the primary subject of this section is the conventional hearing aid, defined as an acoustic amplifier with variable gain and frequency response, several other types of aid are also discussed. In general, we have attempted to include here all types of aids other than electrical cochlear stimulation and sensory-substitution aids.

[2] Further detail on consonant (C) and vowel (V) confusions by cochlear implant subjects can be found in the following original sources: 3M House (C/V: Edgerton *et al.*, 1983a; C: Edgerton *et al.*, 1983b; C/V: Tyler *et al.*, 1985); 3M Vienna (C: Hochmair-Desoyer *et al.*, 1980; V: Hochmair-Desoyer *et al.*, 1981; C: Hochmair and Hochmair-Desoyer, 1983; C/V: Tyler *et al.*, 1985); Nucleus (C: Dowell *et al.*, 1982; C/V: Tyler *et al.*, 1984b; C: Dowell *et al.*, 1985b).

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This report has been reviewed by a group other than the authors according to procedures approved by a Report Review Committee consisting of members of the National Academy of Sciences, the National Academy of Engineering, and the Institute of Medicine.

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